USE OF TEKSCAN PRESSURE SENSORS FOR MEASURING CONTACT PRESSURES IN THE HUMAN KNEE JOINT

S. Herregodts[1], P. De Baets[2], J. Victor[3], and M.A. Verstraete[3]

[1] Ghent University, Belgium
[2] Ghent University, Soete Laboratory, Belgium
[3] Ghent University, Department of Physical Medicine and Orthopaedic Surgery, Belgium

Abstract: The Tekscan pressure sensor is a common instrument to quantify in vitro tibiofemoral and patellofemoral contact pressures, which helps to understand the impact of surgical intervention such as total knee arthroplasty (TKA). As a result of the non-linear behavior of the sensor, the conditioning, normalization and calibration of the sensor are critical to achieve correct measurements. In this paper, a literature review is presented that provides insight in the correct use of these sensors, resulting in optimal accuracy. To guarantee the repeatability of the measurements, a secure and correct fixation of the sensor in the joint is required. Using the sensor for intra-articular measurements induces several unintended effects, which potentially lower the accuracy of the measurement. First, the uneven surface can result in wrinkling and destruction of the sensor, in turn leading to measurement results that can be corrupted. Second, the presence of shear forces on the sensor can lead to wear of the sensor and reduction in sensitivity with loss of accuracy as a result. Also the fixation method can worsen the accuracy. In literature, a deterioration of the accuracy from 3% under optimal conditions to errors of more than 50% are reported as a result of the aforementioned effects.

Keywords: Tekscan, contact pressure, knee joint, K-scan, tibiofemoral, patellofemoral

1 INTRODUCTION

To investigate the influence of surgical techniques such as total knee arthroplasty (TKA) on the human knee joint, quantification of the in vitro tibiofemoral and patellofemoral contact pressure distribution before and after the intervention are an important measure to evaluate the success of the surgery [1–4]. In this paper, the use of the popular K-scan pressure sensor from Tekscan (South Boston USA) for in vitro pressure measurements in the human knee joints is discussed. This sensor is constructed by two thin mylar layers, onto which electrical conductive leads have been deposited. The two layers are bonded together with a layer of semiconductive ink. At the intersection points of the conductive leads, pressure sensitive cells are constructed (see Figure 1). The electrical resistance of each cell is dependent of the force acting on that cell. By measuring the resistance of all cells, the pressure distribution on the sensor can be derived after an appropriate calibration [5, 6] . To measure patellofemoral contact pressure, a sensor with measuring area of 55,9mm x 55,9mm and sensel density of 62/cm^2 is commonly used. Tibiofemoral contact pressure is usually measured with a sensor which has two separate measuring pads of 27,9mm x 33,0mm and sensel density of 62/cm^2.

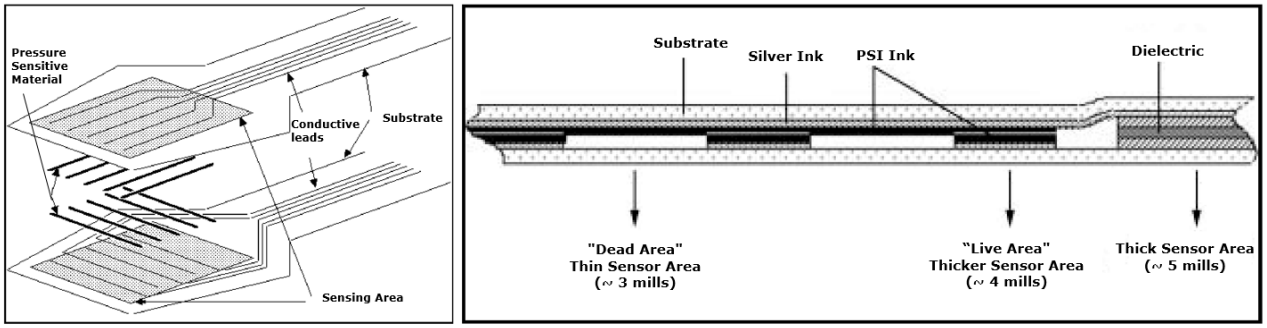

Figure 1: Construction Tekscan sensor [7]

The success of this sensor can be attributed to the low thickness (0,1mm) and the real-time evaluation of contact pressures. As a result, this sensor is well suited to evaluate contact pressures during dynamic in vitro tests [5, 8–10]. Despite these favourable characteristics, there are also quite some drawbacks when

this sensor is used to measure intra-articular contact pressures. These drawbacks are discussed in the remaining sections.

2 SENSOR CALIBRATION

It has been shown that this sensor exhibit several forms of nonlinear behaviour. As a result, the calibration procedure has a fundamental impact on the final accuracy. The resistance of the electrical cells has a semi hyperbolic behaviour with respect to the applied force. The measuring device internally converts this to a semi exponential function, which represents the digital raw value of the force with respect to the applied load. This digital raw value forms the basic output, which has to be calibrated to achieve a relationship between contact pressure and the digital raw value [7]. To minimize the influences of hysteresis, drift and to improve the repeatability, the sensor should be conditioned. This preconditioning can be done by repeatedly applying a homogenous pressure exceeding the testing force [7, 11].

It should be noted that each cell has a different resistance-force response. To overcome this unwanted characteristic, the sensor should be equilibrated prior to calibration. Equilibration is a normalization process of the output of each cell to the average output value of the whole pad. Therefore the sensor pad should be loaded by a homogeneous pressure. This can be achieved by the use of a pneumatic or hydraulic bladder or by loading it between parallel steel plates while the sensor is sandwiched between two rubbery sheets with a low poison ratio. To achieve a homogenous sensor sensitivity over the full measuring range, this equilibration can be done at different pressure levels [7].

Following equilibration, the sensor can be calibrated. As the sensor's sensitivity is temperature dependent, equilibration and calibration should be done at the temperature the sensor will be used later on. The sensor is composed of a matrix of sensing elements surrounded with dead zones (see Figure 1). The stiffness of these dead zones and the measuring cells is different, consequently the interfacing material and surface structure will have an influence on the measurement. Therefore it is very important that the surfaces used for calibration mimic the properties of the testing surfaces as good as possible.

When testing materials with high poison ration or in situations where shear forces could be transmitted to the sensor, the use of a lubricant or a low friction protection film, such as Teflon, is advised. These measures will lower the shear forces acting on the sensor which could damage the sensor or corrupt the measurement data [7, 11]. As this sensor exhibits time dependent drift behaviour, the calibration time should be chosen close to the testing time [7, 12, 13]. Two calibration methods are provided by Tekscan. The first one is a linear single point calibration. This method is only satisfying for measurements within a small pressure range. The second method is the power law calibration. In this method a power curve is fitted. A minimum of two calibration points should be used for fitting this curve. To achieve a satisfactory overall calibration, however, up to 10 calibration points can be used. As the digital raw output has a quasi power law behaviour, this method provides an acceptable full range calibration. It has however been demonstrated that the use of a customized 3 and 10-point polynomial calibration improves the overall accuracy furthermore from 2,7% for a power law calibration to respectively 1,2% and 0,6 % [14, 15]. Another study showed an increased accuracy from 14% for 3 point power law calibration to 4% for a customized 11 point calibration [16].

3 APPLICATION IN KNEE BIOMECHANICS

3.1 Mounting techniques

When actuating the knee in a knee simulator, the sensor should be securely fixed to one of the two contact surfaces to obtain accurate and repeatable measurements. To measure the patellofemoral contact pressure distribution during in vitro tests with native knees, the Tekscan sensor is mostly fixed to the patella (see Figure 2) [17–19], only one experiment in which the sensor was fixed to the trochlea of the femur was reported [20]. If the tibiofemoral contact pressure has to be measured, the sensor can be fixed on the femoral plateau underneath [21–25] or on top of the menisci [26]. Also complete removal of the menisci to install the sensors has been reported [27]. The sensor can be fixed by the use of sutures (see Figure 2) [17, 18, 23, 27–29] or by metal pins [30, 31]. Alternatively, the use of a suture-glue combination has been reported [4].

Figure 2: K-scan 4000 sensor attached to the patella by sutures [18]

For testing with a knee prosthesis, a common practice is to glue the sensor onto the patella or tibial tray [32], or fixed by the use of double sided tape [2, 33]. To minimize shear forces acting on the sensor, a protective Teflon film of 0,1mm is often applied [4, 17–19, 21, 22, 28, 29, 33]. Alternatively, a lubricant such as surgical lubricant [5] or vaseline [22] can be applied onto the sensor.

3.2 Sensor behaviour in human knee joint

One of the first problems that arise when using the Tekscan sensor for intra-articular contact pressure measurements is wrinkling of the sensor [5, 7, 10, 34–36]. As the sensor is made of a dimensionally stable substrate, it can nicely roll around a cylinder but it is difficult to wrap it around a ball shaped surface. Due to the high stiffness of the sensor, bending in two directions is impossible [10]. As the global shape of the patella is convex, the sensor cannot tightly fit to the complete surface (see Figure 2). By wrinkling of the sensor, local inexistent high pressures will be read or cells could break down. A much lower repeatability was found for measurements on curved surfaces compared to flat surfaces [5].

By testing a knee in a knee simulator, the contacting surfaces will undergo a sliding motion. As a result the sensor will be exposed to shear forces. The manufacturer advices to avoid shear forces acting on the sensor as these reduce the lifetime and alter the performance of the sensor [7]. If shear forces can be expected, measures should be taken to avoid the transmitting of these shear forces onto the sensor. This can be done by applying a sheet with low friction coefficient like Teflon on the sensor or by lubricating the sensor.

When exposing the sensor to a humid environment like in a knee, the sensor sensitivity will diminish over time. This can successfully be solved by sealing the sensor by the use of a plastic cover [37] or by saturating the sensor prior to calibration and testing [38].

3.3 Accuracy of Tekscan sensor for intra-articular pressure measurements

The accuracy of the Tekscan sensor changes when used for intra-articular measurements compared to static measurements between flat surfaces [5, 19, 29, 37]. However, most researchers use the accuracy of a static calibration test between flat surfaces without performing a validation test to define the accuracy level of the intra-articular pressure measurements. Several studies have shown that this is not correct [5, 29]. A large variation in accuracy of the sensor is reported. This is a result of the difference in the conditioning, equilibration, calibration and testing process. For flat compression tests, an accuracy of 3% for the applied force and 9% for the applied pressure was observed by Kent et al. [8]. A much worse accuracy for a flat pressure test between −12% and +20% was reported by Martinelli et al. [10]. For patellofemoral contact measurements, Wilson et al. [5] reported a force accuracy of 6,5±4,4% (mean±standard deviation over five trials each at seven load levels). The standard deviation for patellofemoral contact force measurement was assessed at 9,1% by Wilson at al. [5]. The correctness of these intra-articular measurement accuracies can be questioned as several reported causes of measurement errors are probably not taken into account. Next to the change in contact mechanics of the joint by inserting the sensor, four major causes of sensor accuracy deterioration for intra-articular measurements are reported. The first one is the effect of the sensor fixation. The second one is the effect of a humid environment. The third one is the effect of wrinkling of the sensor as a result of placing it between uneven surfaces. The last potential cause is the presence of shear forces acting on the sensor. These causes are discussed in more detail in this section. Note furthermore that shear forces and sensor wrinkling are the primary lifetime determining factors of this sensor [7, 29].

Inserting the sensor film in a biological joint can change the contact mechanics. Inserting a 0,3mm thick Fuji Prescale pressure film with an elastic modulus 100–300 times larger than that of articular cartilage can distort contact pressure measurements by 10–26% [37]. As the Tekscan sensor has a lower thickness, a lower effect can be expected but results have not been published to the authors' knowledge.

Little research has been done to investigate the influence of the mounting method of the sensor on the accuracy of the measurements but several researchers have questioned this [33]. Only the influence of cementing the sensor has been investigated. By cementing the sensor, the resultant force was found to be 31±32% lower and the contact area to be 24±13% lower in comparison with an uncemented sensor. The maximum pressure on the other hand was found to be 24±44% higher [5]. This discrepancy could be a result of stiffening the contact.

Exposing the sensor to the humidity of a knee can lower the sensor output by 25% after 6 hours [38]. By saturating the sensor prior to calibration or by shielding the sensor this can be counteracted.

The effect of sensor wrinkling has not yet been quantified, but false pressure "spikes" where observed in the measurements of Ostermeier et al. [19]. In addition, broken cells were observed after testing [36]. Apart from the visible sensor wrinkling, the effect of using the sensor between curved surfaces such as the patellofemoral contact is not clear and can probably not be uniquely quantified [5].

Shear forces acting on the sensor are often reported as a plausibly cause for contact pressure measurement errors but were not profoundly investigated [39, 40]. Wilharm et al. (2013) [29] were the first to investigate and report the effects of shear forces on the sensor that occur when the knee is tested in a knee simulator. Using the sensor without a Teflon protection film resulted in destruction of the sensor after 1 cycle of knee flexion. With the use of a protective film, a reduction in output of 17.2% after 12 cycles was observed. This effect could be attributed to sensor deterioration as a result of wear by shear forces. As the deterioration of the sensor seems to be linear, this systematic error can be quantified and compensated [29].

Based on the results from literature, a workflow for performing correct intra-articular pressure measurements with Tekscan sensors is presented in Figure 3. It is important to check the sensor for wear and failure after performing the measurements to estimate errors caused by sensor wear. To determine the final accuracy of the dynamic intra-articular pressure measurements, a validation test should be performed, which mimics the intra-articular situation and thereby takes into account the effects of shear force, uneven surfaces, wrinkling and the fixation method. In addition, the effect of the time dependent drift should be characterized. During dynamic testing the pressures are fluctuating, hence neither a single testing time nor a constant force can be defined.

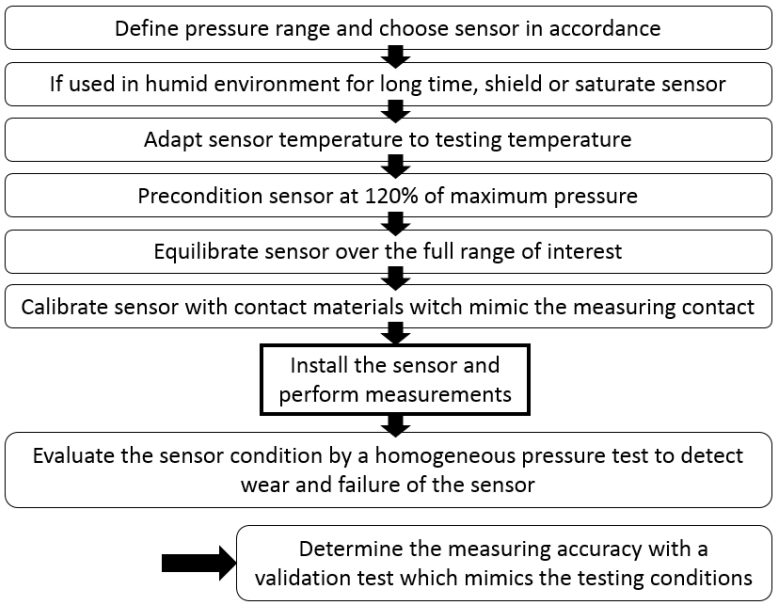

Figure 3: Workflow for performing correct intra-articular pressure measurements with Tekscan sensors

4 CONCLUSIONS

The Tekscan sensor has proven to be an interesting sensor for dynamic measurements of patellofemoral and tibiofemoral contact pressures. Its low thickness and the continuous measurements are the major advantages. As this sensor exhibits several forms of non-linear behaviour, the calibration method has a large effect on the accuracy of the sensor. The basic reported accuracy for flat static pressure measurements varies from 3% till 20%. The large variation is a result of the difference in the conditioning, equilibration, calibration and testing process. By using the sensor for intra-articular pressure measurements, effects from humidity, shear forces, wrinkling of the sensor, uneven test surfaces and the fixation methods can affect the final accuracy. Few research has yet been performed to quantify these effects but errors of 31±32% when cementing the sensor and 17,2% when exposed to shear forces have been reported. It is clear that care must be taken when using this sensor for intra-articular pressure measurements or errors of over 50% can occur. A validation test that mimics the testing conditions should be included to determine the final accuracy of the pressure measurements.

5 REFERENCES

1. Becher C, Huber R, Thermann H, Paessler HH, Skrbensky G (2008) Effects of a contoured articular prosthetic device on tibiofemoral peak contact pressure: a biomechanical study. Knee Surgery, Sports Traumatology, Arthroscopy 16:56–63

2. Harris M, Morberg P, Bruce W, Walsh W (1999) An improved method for measuring tibiofemoral contact areas in total knee arthroplasty: a comparison of K-scan sensor and Fuji film. Journal of biomechanics 32:951–958

3. Jason K, Ryuichi G, Michelle H, and Thay Q (2012) Effects of Kneeling on Tibiofemoral Contact Pressure and Area in Posterior Cruciate-Retaining and Posterior Cruciate-Sacrificing Total Knee Arthroplasty. The Journal of Arthroplasty 27:

4. Becher C, Heyse TJ, Kron N, Ostermeier S, Hurschler C, Schofer MD, Fuchs-Winkelmann S, Tibesku CO (2009) Posterior stabilized TKA reduce patellofemoral contact pressure compared with cruciate retaining TKA in vitro. Knee Surgery, Sports Traumatology, Arthroscopy 17:1159–1165

5. Wilson D, Apreleva MV, Eichler MJ, Harrold FR (2003) Accuracy and repeatability of a pressure measurement system in the patellofemoral joint. Journal of biomechanics 36:1909–1915

6. Tekscan (2003) Tekscan: Medical sensor catalog.

7. Anselmi R (2003) I-Scan Equilibration and Calibration Practical Suggestions.

8. Bachus KN, DeMarco AL, Judd KT, Horwitz DS, Brodke DS (2006) Measuring contact area, force, and pressure for bioengineering applications: using Fuji Film and TekScan systems. Medical engineering and physics 28:483–488

9. Giacomozzi C (2010) Appropriateness of plantar pressure measurement devices: A comparative technical assessment. Gait and posture 32:141–144

10. Martinelli L, Hurschler C, Rosenbaum D (2006) Comparison of capacitive versus resistive joint contact stress sensors. Clinical orthopaedics and related research 447:214–220

11. Tekscan (2003) Tekscan I-Scan System Start-Up.

12. Papaioannou G, Protopappas VC, Panagiotis T, Mitrogiannis C, Nianios G, Tashman S, Papaioannou G (2008) A new method for pressure sensor equilibration and conditioning. Brazilian Journal of Biomotricity 2:176–195

13. Otto J, Brown T, Callaghan J (1999) Static and dynamic response of a multiplexed-array piezoresistive contact sensor. Experimental Mechanics 39:317–323

14. Brimacombe JM, Wilson DR, Hodgson AJ, Ho KC, Anglin C (2009) Effect of calibration method on Tekscan sensor accuracy. Journal of biomechanical engineering 131:034503

15. Brimacombe M, Anglin C, J. H, Wilson R (2009) Validation of calibration techniques for tekscan pressure sensors. ASB 29th Annual Meeting

16. DeMarco A, Rust D, Bachus K (2000) Measuring contact pressure and contact area in orthopedic applications: Fuji film vs. Tekscan. Trans Orthop Res Soc 25:518

17. Steinbrück A, Schröder C, Woiczinski M, Fottner A, Müller PE, Jansson V (2013) Patellofemoral contact patterns before and after total knee arthroplasty: an in vitro measurement. Biomedical engineering online 12:58

18. Heyse TJ, Becher C, Kron N, Ostermeier S, Hurschler C, Schofer MD, Tibesku CO, Fuchs-Winkelmann S (2010) Patellofemoral pressure after TKA in vitro: highly conforming vs. posterior stabilized inlays. Archives of orthopaedic and trauma surgery 130:191–196

19. Ostermeier S, Holst M, Bohnsack M, Hurschler C, Stukenborg-Colsman C, Wirth C-J (2007) Dynamic measurement of patellofemoral contact pressure following reconstruction of the medial patellofemoral ligament: An in vitro study. Clinical Biomechanics 22:327–335

20. Bicos J, Carofino B, M. A, Schepsis A, Fulkerson J, Mazzocca A (2006) Patellofemoral Forces After Medial Patellofemoral Ligament Reconstruction. The journal of knee surgery 19:

21. Von Lewinski G, Hurschler C, Allmann C, Wirth C (2006) The influence of pre-tensioning of meniscal transplants on the tibiofemoral contact area. Knee Surgery, Sports Traumatology, Arthroscopy 14:425–436

22. Von Lewinski G, Stukenborg-Colsman C, Ostermeier S, Hurschler C (2006) Experimental measurement of tibiofemoral contact area in a meniscectomized ovine model using a resistive pressure measuring sensor. Annals of biomedical engineering 34:1607–1614

23. Padalecki JR, Jansson KS, Smith SD, Dornan GJ, Pierce CM, Wijdicks CA, LaPrade RF (2014) Biomechanical Consequences of a Complete Radial Tear Adjacent to the Medial Meniscus Posterior Root Attachment Site In Situ Pull-out Repair Restores Derangement of Joint Mechanics. The American journal of sports medicine 42:699–707

24. Ode GE, Van Thiel GS, McArthur SA, Dishkin-Paset J, Leurgans SE, Shewman EF, Wang VM, Cole BJ (2012) Effects of serial sectioning and repair of radial tears in the lateral meniscus. The American journal of sports medicine 40:1863–1870

25. Paci JM, Scuderi MG, Werner FW, Sutton LG, Rosenbaum PF, John P (2009) Knee Medial Compartment Contact Pressure Increases With Release of the Type I Anterior Intermeniscal Ligament. Medicine The American Journal of Sports

26. Guettler JH, Demetropoulos CK, Yang KH, Jurist KA (2004) Osteochondral Defects in the Human Knee Influence of Defect Size on Cartilage Rim Stress and Load Redistribution to Surrounding Cartilage. The American journal of sports medicine 32:1451–1458

27. Gadikota HR, Kikuta S, Qi W, Nolan D, Gill IV TJ, Li G (2013) Effect of Increased Iliotibial Band Load on Tibiofemoral Kinematics and Force Distributions: A Direct Measurement in Cadaveric Knees. journal of orthopaedic & sports physical therapy 43:478–485

28. Bohnsack M, Klages P, Hurschler C, Halcour A, Wilharm A, Ostermeier S, Rühmann O, Wirth C-J (2009) Influence of an infrapatellar fat pad edema on patellofemoral biomechanics and knee kinematics: a possible relation to the anterior knee pain syndrome. Archives of orthopaedic and trauma surgery 129:1025–1030

29. Wilharm A, Hurschler C, Dermitas T, Bohnsack M (2013) Use of Tekscan K-Scan Sensors for Retropatellar Pressure Measurement Avoiding Errors during Implantation and the Effects of Shear Forces on the Measurement Precision. BioMed research international 2013:

30. Kock NB, Smolders JM, Van Susante JL, Buma P, Van Kampen A, Verdonschot N (2008) A cadaveric analysis of contact stress restoration after osteochondral transplantation of a cylindrical cartilage defect. Knee Surgery, Sports Traumatology, Arthroscopy 16:461–468

31. Knupp M, Stufkens SA, van Bergen CJ, Blankevoort L, Bolliger L, Van Dijk C, Hintermann B (2011) Effect of supramalleolar varus and valgus deformities on the tibiotalar joint: a cadaveric study. Foot \& ankle international 32:609–615

32. Matsuda S, White SE, Williams II VG, McCarthy DS, Whiteside LA (1998) Contact stress analysis in meniscal bearing total knee arthroplasty. The Journal of arthroplasty 13:699–706

33. Arnout N, Vanlommel L, Vanlommel J, Luyckx J, Labey L, Innocenti B, Victor J, Bellemans J (2014) Post-cam mechanics and tibiofemoral kinematics: a dynamic in vitro analysis of eight posterior-stabilized total knee designs. Knee Surgery, Sports Traumatology, Arthroscopy 1–11

34. Anselmi R (2013) I-Scan® & High Speed I-Scan® User Manual.

35. Elguizaoui S, Flanigan DC, Harris JD, Parsons E, Litsky AS, Siston RA (2012) Proud osteochondral autograft versus synthetic plugs-Contact pressures with cyclical loading in a bovine knee model. The Knee 19:812–817

36. Yamamoto A, Massimini DF, DiStefano J, Higgins LD (2014) Glenohumeral Contact Pressure With Simulated Anterior Labral and Osseous Defects in Cadaveric Shoulders Before and After Soft Tissue Repair. The American journal of sports medicine 42:

37. Wu J, Herzog W, Epstein M (1998) Effects of inserting a pressensor film into articular joints on the actual contact mechanics. Journal of biomechanical engineering 120:655–659

38. Jansson KS, Michalski MP, Smith SD, LaPrade RF, Wijdicks CA (2013) Tekscan pressure sensor output changes in the presence of liquid exposure. Journal of biomechanics 46:612–614

39. Latt LD, Glisson RR, Montijo HE, Usuelli FG, Easley ME (2011) Effect of graft height mismatch on contact pressures with osteochondral grafting of the talus. The American journal of sports medicine 39:2662–2669

40. Becker R, Wirz D, Wolf C, Göpfert B, Nebelung W, Friederich N (2005) Measurement of meniscofemoral contact pressure after repair of bucket-handle tears with biodegradable implants. Archives of orthopaedic and trauma surgery 125:254–260